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Laser Assisted Vascular Welding With Real Time Temperature Control

Robert B. Stewart, PhD, Aziz Benbrahim, MD, Glenn M. LaMuraglia, MD, Meir Rosenberg, MS, Gilbert J. L'Italien, PhD, William M. Abbott, MD, and Robert T.V. Kung, PhD

ABIOMED, Inc., Danvers, Massachusetts 01923 (R.B.S., M.R., R.T.V.K.); Division of Vascular Surgery, Massachusetts General Hospital, Harvard Medical School, Boston, Massachusetts 02114 (A.B., G.M.L., G.J.L., W.M.A.)

Background and Objective: **Previous studies in laser assisted vascular welding have been limited by the lack of a reliable end point for tissue fusion. As a means of improving the reproductibility of laser assisted repairs, a system incorporating real time temperature monitoring and closed loop feedback was used.**
Study Design/Materials and Methods: **The system consisted of a direct view infrared thermometer for monitoring the laser heated spot, a 1.9 μm diode laser, and a microprocessor for data acquisition and feedback control of the laser power to maintain a constant tissue temperature. Rat aortas were welded under constant surface temperature conditions.**
Results: **In vivo temperature stability of ±2°C was achieved over a temperature range of 70–90°C pertinent to welding small vessels. When welds were completed using the feedback system to maintain the tissue temperature at 80°C, the acute success rate was 100% and the burst pressure was 290 ± 70 mmHg.**
Conclusion: **These studies demonstrate that the use of real time monitoring and feedback control results in improved consistency for vascular tissue welding.**

Key words: diode laser, feedback loop, infrared thermometry, real time control, thermal weld

INTRODUCTION

Although laser welding of blood vessels was first demonstrated over 15 years ago [1], only a limited number of clinical studies of laser use for vascular and microvascular anastomoses have been completed [2,3]. This lack of interest is somewhat puzzling when the surgical benefits of the laser procedure are considered. These benefits include shorter operative times, reduced foreign body reaction, reduced bleeding, and technical ease of use for small caliber vessels. The technical ease of the laser procedure has additional advantages for any minimally invasive vascular procedures, where the conventional suture-tying technique is extremely difficult.

The main disadvantage of the laser assisted procedure is the low strength of the resulting anastomosis, especially in the acute healing phase up to 4 days postoperatively. This can result in an anastomotic aneurysm, or even rupture of the repair. Another perceived disadvantage of laser tissue welding is inconsistent results between different research groups/surgeons for similar laser procedures. This indicates that the surgical technique is either not reproducible or not obvious to a highly trained surgeon. It is also noteworthy that the low initial anastomotic strength may also be indicative of technical variations during a single procedure.

In a typical laser assisted anastomosis procedure, the surgeon looks for a subtle visible change in the tissue, either blanching or some

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Address reprint requests to Robert B. Stewart, ABIOMED, Inc., 24 Cherry Hill Drive, Danvers, MA 01923.

discoloration of the tissue surface, as an ad hoc end point for completion of the weld. This is clearly a highly subjective and qualitative indicator. As a means of quantifying changes during tissue welding, several techniques have been suggested for real time monitoring.

Tissue parameters that may provide diagnostic information for tissue welding include native autofluorescence [4], optical birefringence [5], and the temperature of the tissue [6]. Prior studies in tissue welding indicate that the laser acts solely as a source of thermal energy and no photochemical processes are important. Therefore, a simple Arrhenius model should be valid for the tissue welding reaction rate. For this model, the extent of the reaction is linearly proportional to time and exponential in temperature. As a result of this exponential dependence, temperature is the dominant parameter for governing the extent of the reaction as illustrated in the following example. The activation energy of tissue denaturation is known to be about 100 kcal/mole [7]. For a ±2°C variation in tissue temperature around 80°C (from 78 to 82°C), the reaction rate varies by a factor of 5. Analogously, the welding process should be sensitive to the tissue temperature, which should provide an excellent real time monitor for the laser welding procedure.

By monitoring the tissue temperature during the welding process, the optimal temperature range for laser assisted anastomoses can be determined. The optimal temperature range is defined as the range that gives the most desirable clinical outcome. The clinical parameters of primary importance for vascular welding are the acute success rate and the anastomotic strength. For successful repairs, the optimal temperature range can be determined by correlating acute burst strengths with the tissue temperature during welding. In addition, once the optimal temperature range is known, a feedback loop can be employed to modulate the laser power to maintain the tissue temperature within this range during tissue welding. This should result in a reproducible, reliable laser assisted anastomosis.

We have constructed a laser/infrared thermometer system capable of controlling the surface temperature to ±2°C in situ during tissue welding. The controlled surface temperature was maintained during in vivo welding, and the integrity of the resulting repair was determined. For successful welds, acute burst pressure measurements were completed.

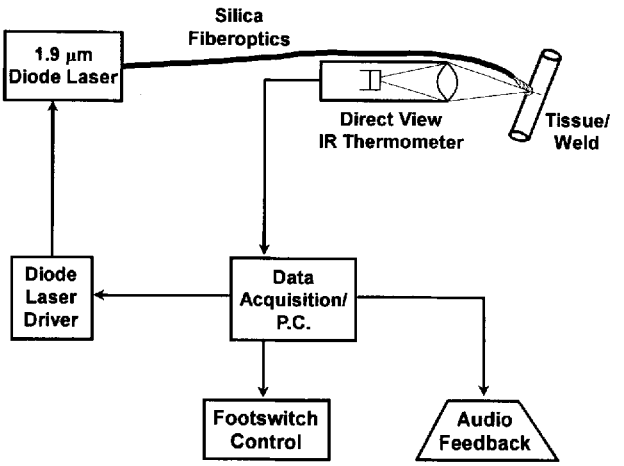

Fig. 1. Block diagram for laser/infrared thermometer system used for thermally controlled vascular welding.

MATERIALS AND METHODS

The acute studies reported here were designed to determine the optimal temperature range for consistent small blood vessel welding.

Laser/Infrared Thermometer System

A diagram of the laser/infrared thermometer system is given in Figure 1. The system consists of three parts: a fiber optic coupled 1.9 μm diode laser for power delivery to the tissue, an infrared thermometer for tissue surface temperature measurements, and a microprocessor based system for temperature data acquisition and real time laser power control to maintain a constant surface temperature. In addition, an audio feedback system is used to prompt the surgeon when the desired weld temperature has been reached. Details of each of the system components are given later.

The laser used is a low power 1.95 μm diode laser [Spectra Diode Laboratories (SDL)], driven by a commercially available laser diode driver (SDL model SDL-820). This laser was chosen because the tissue absorption depth, defined as the 1/e-fold absorption length, at this wavelength closely matches the vessel wall thickness for small diameter vessels. This matching results in uniform deposition of energy across the thickness of the weld [8]. For these welding studies, the laser power was limited to 300 mW, as delivered to the tissue via a 300 μm (core diameter) silica fiber. The advantages of using a diode laser in our system include high electrical to optical efficiency, high device reliability, and rapid power modulation via feedback control of the diode cur-

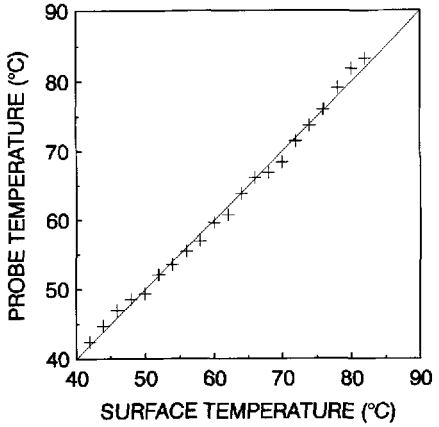

Fig. 2. Calibration for the infrared thermometer system. The temperature of the probe used for thermally controlling the weld is plotted against the known surface temperature, which was measured with a thermocouple and a commercially available infrared thermometer (average value shown).

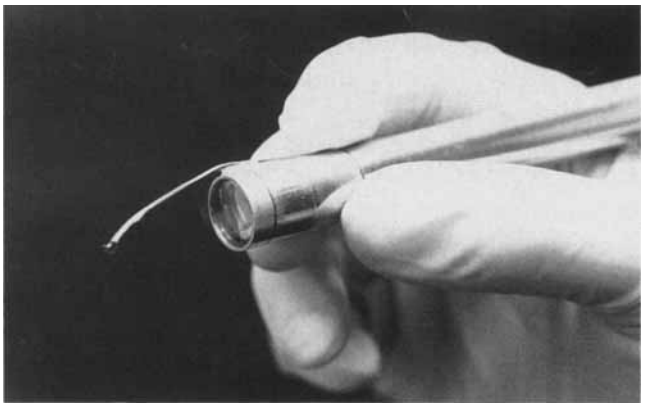

Fig. 3. Photograph of the handpiece used for the welding studies. This handpiece incorporates the laser fiber and the infrared thermometer.

rent. The time response of the diode power was limited by the driver to 100 kHz.

The infrared thermometer used was a direct viewing device, which monitored a 0.4 mm spot, as measured by scanning a knife edge across the image plane in the laser heated region. This spot was imaged directly onto a thermopile (Heimann GmbH model TPS 434-T) using a single ZnSe lens. Using these two elements for the direct view thermometer allowed imaging radiation over a range of 8–13 μm. Owing to the strong absorption of water over this range and the high water content in vascular tissue, a vessel can be considered to be a blackbody with an emissivity of 0.99 for the temperature measurement. In addition, since the penetration depth of radiation in the 8–13 μm range is very short (about 10 μm), the infrared thermometer employed here only viewed the surface temperature of the tissue. The thermometer was calibrated using a resistively heated black plate. The temperature of this plate was measured using a thermocouple *and* another infrared thermometer (Omega Instruments OS-88000-K-1200), which viewed the same region of the black plate as our thermometer. These two independent measurements agreed to ±0.25°C. The calibration curve obtained for our infrared thermometer system vs. the average of the other two measurements is given in Figure 2. Based on this curve, the temperature measurement is accurate to ±1°C.

The laser delivery fiber and the infrared thermometer were incorporated into a hand-held surgical instrument. A photograph of this handpiece is provided in Figure 3. A stainless steel tube is used to direct the fiber to the weld region, and a wire guide attached to this tube is used to define the welding region and to provide tactile feedback for the surgeon. The infrared thermometer, consisting of the thermopile and the imaging lens, are enclosed in the body of the handle. This arrangement resulted in a 0.7 mm laser heated spot concentric with a 0.4 mm spot viewed by the thermometer. Although the laser spot size was approximately twice that used in previous studies by other groups [9], past studies using a 1.9 μm laser have demonstrated good results with this spot size [10].

The output from the infrared thermometer can be processed further using either a personal computer (PC) based system, or by using a dedicated microprocessor for data acquisition and control. For the studies described later, a PC based system was used with a data acquisition/control card and its corresponding software. This allowed easy storing and plotting of the resulting thermal profiles obtained while tissue welding. A simple feedback loop was used, which changed the laser current every 5 ms based on both the difference between the actual temperature and the desired temperature, and the difference in the actual temperatures for successive measurements. The actual laser control was accomplished via an analog input on the current driver.

Surgical Procedure

Twelve CD Fisher rats (250–300 grams) were used. The protocol used was approved by the Institutional Animal Care Committee. All ani-

mals used in the study received humane care in compliance with the *Principles of Care and Use of Laboratory Animals* prepared by the National Academy of Sciences and published by the National Institutes of Health (NIH publication no. 80-23, revised 1985). The animals were anesthetized with ketamine (35 mg/ml) and xylazine (5 mg/ml). The abdominal aorta was isolated via a laparotomy using standard microsurgical techniques. A 1 mm transverse arteriotomy was made using a scalpel. This repair model allows a more controlled weld than a complete end-to-end anastomosis, but still requires all of the major elements (e.g., precise tissue apposition) used in an end-to-end repair. The open vessel was irrigated with heparinized saline (120 μg/ml). A 10-0 nylon stay suture was placed at each end of the arteriotomy (just beyond the incision) to provide traction.

The laser weld was completed under microscopic viewing. While traction was applied to the two stay sutures, the wire guide on the probe was placed against the tissue surface at one end of the arteriotomy. The laser/control system was activated via the footswitch, and the tissue temperature rose to the desired set temperature while the surgeon held the probe in place. Until the desired temperature was achieved, an audio alarm signaled the surgeon to hold the probe in place. Once the tissue was at the desired temperature, the audio alarm ended, signaling the surgeon to advance the probe along the tissue surface. The surgeon completed the weld by moving the fiber across the weld in a continuous fashion. The laser power was automatically varied during the anastomosis to maintain the desired surface tissue temperature. The total weld time was approximately 5–8 s for the 1 mm long weld. No active measures were taken to limit the exposure time.

In order to minimize the number of rats needed, two laser welds were completed on each rat aorta in a distal to proximal fashion. This progression permitted multiple burst pressure measurements via a femoral artery pressure line.

Weld Success Rate and Burst Pressure Tests

Following the laser procedure, the resultant weld integrity was tested. The first test for acute success was the ability of the weld to withstand systemic pressure. Next, the standard empty and fill patency test was completed. A third patency test was completed by monitoring the femoral line pressure. Observation of a physiologic pulsatile pressure was an indication of a patent aorta.

Once a repair was determined to be successful, burst pressure measurements were completed. The femoral pressure line tubing was advanced to directly distal to the repair. The other end of this tubing was connected to a calibrated pressure transducer and a Harvard infusion pump. Intraluminal pressure was continuously monitored on a strip chart recorder. The vessel was cleansed of blood by gentle irrigation with heparinized saline, while a proximal clamp was used to isolate the vessel. Burst pressure measurements were carried out by infusing heparinized saline at a rate of about 0.3 ml/s. The burst pressure was recorded as that pressure at which a stream of saline was viewed visually leaking at the anastomosis. This event is accompanied by an immediate fall in the intraluminal pressure. The burst pressure was defined as the maximum pressure noted before the fall in the intraluminal pressure. As mentioned earlier, a second weld was completed for each animal. This was accomplished by advancing the femoral line proximally past the first repair and ligating the vessel below the second arteriotomy. This allowed the second weld to be tested under physiologic conditions.

No histologic evaluation of the burst specimens was completed. Although histology may provide information on the extent of thermal injury produced at different temperatures, the studies reported here were primarily concerned with the mechanical properties of the acute weld, particularly the acute burst pressure.

RESULTS

A series of rat arteriotomies were repaired using the laser/infrared thermometer system. A sample of the data obtained for a thermally controlled laser weld is given in Figure 4. The system was set to control the tissue temperature at 80°C. As can be seen in Figure 4, the laser heated the tissue to the desired weld temperature in about 1 s. At this point the surgeon began moving the probe across the weld. The surface temperature was maintained via the feedback loop at 80 ± 2°C during the weld. Figure 4 also gives the temperature history for a successful tissue weld that was completed by the surgeon using the traditional visual feedback. These open loop data show a longer temperature rise time, which is indicative of premature translation of the delivery fiber prior to the tissue attaining a constant temperature. Note also that the temperature varied substantially during the open loop weld, even though

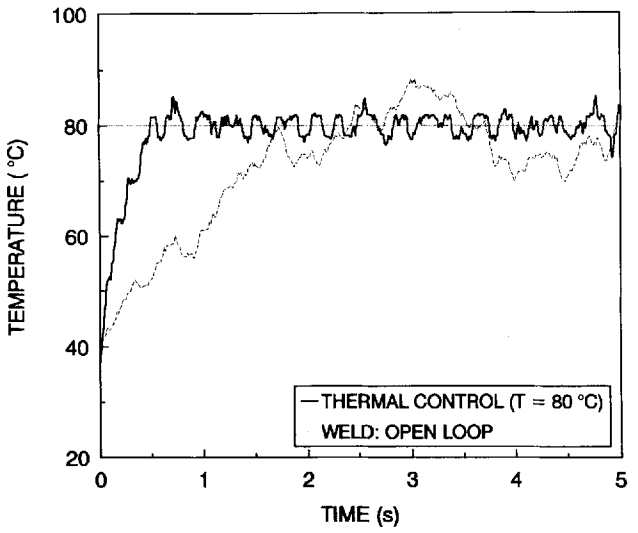

Fig. 4. Temporal profile recorded for the surface temperature during a laser repair of a rat aorta. The laser was footswitch activated at time t = 0 by the surgeon.

the surgeon attempted to maintain a constant visual effect by varying the translation speed of the laser beam across the tissue. Using this open loop control method, the surgeon was able to maintain a surface temperature of 78 ± 5°C for the weld. The surgeon's visual control was taken as the mean and standard deviation of the temperature data given in Figure 4 for time greater than 2 s in the figure. If the data for time greater than 1 s are used, the temperature was controlled to 76 ± 6°C in the open loop case shown.

When thermally controlled welds were completed over a range of temperatures, the results given in Table 1 were obtained. At set surface temperatures below 70°C, no visual tissue welding occurred. Some degree of tissue fusion was achievable over a relatively wide temperature range of 70–90°C. At the lower end of this range (70°C), however, the welds were too fragile to withstand systemic pressure. For the higher temperatures (>90°C), significant tissue shrinkage caused excessive narrowing and sometimes artery occlusion. Successful vessel welding occurred over the range of 75–90°C; however, 100% success rates only occurred within the middle of this range. These results suggest that, although the range of temperatures for which some type of tissue fusion may occur is relatively broad, the temperature range for welds having suitable weld strength is significantly narrower. Furthermore, the range for reproducible welds, defined as having a 100% acute success rate with acceptable burst strengths, is still narrower.

Another comparison can be made between the thermally controlled, or closed loop, weld and the weld completed open loop by the same surgeon, relying on the normal visual cue. The open loop results are also given in Table 1. Note that both the success rate and burst pressures were significantly higher ($P < 0.05$ for the burst pressures) for the closed loop welds controlled near the "optimal" temperature compared with the open loop welds.

Finally, it is noteworthy that these welds were completed by transversing the fiber at approximately 0.2 mm/s as described earlier. A significant increase/decrease in the translation speed would shift the effects described to higher/lower temperatures, as expected for an Arrhenius process. However, the welding kinetics should be much less sensitive to the exposure time vs. the temperature, since the reaction is only linear with time, as opposed to being exponential with temperature.

DISCUSSION

Thermally controlled laser assisted tissue welding represents a new type of surgical procedure, in which real time diagnostics are used with feedback loops to control the surgical outcome. By using feedback loops, decisions can be made at rates much faster than the surgeon's reaction time. In our system, the laser power was adjusted every 5 ms to maintain the tissue surface at a nearly constant temperature. In the traditional open loop vessel welding procedure, the surgeon "controls" the fusion process by changes in the delivery rate, which is the rate at which the delivery fiber traverses the tissue. It is doubtful that the surgeon could compensate at rates above 10 Hz when relying on visual cues alone. In addition, inhomogeneities in tissue properties and other variables are less problematic for a more temporally "agile" closed loop system.

It is noteworthy that the rather crude traditional open loop end points, such as tissue discoloration or blanching, provided adequate feedback for tissue welding. This may be due to the fact that the temperature range over which some type of tissue fusion occurs is relatively broad (see Table 1), similar to what other groups have reported for thermal welding in vivo [11]. However, the substantially narrowed range for consistent successful vessel welding (i.e., having a 100% success

TABLE 1. Success Rates and Burst Pressures for Laser Assisted Vascular Welding Using Real Time Temperature Control

Temperature (°C)	N	Acute success (%)	Burst pressure mmHg (SD*)
70	4	0	—
75	3	66	310 (50)
80	5	100	290 (70)
90	4	75	300 (50)
95	2	0	—
Open loop	6	66	200 (90)

*SD = standard deviation.

rate and high burst strengths), may explain the inconsistent results between different research groups. Whereas each group may have achieved successful tissue welds, the exact point of the tissue fusion, as estimated by visual cues, could have differed significantly so that one group was working near the lower temperature threshold, while another group might be working near the upper bound. In fact, the terms *overcooking* and *undercooking*, commonly used by practitioners of tissue welding to describe laser-tissue effects, are further evidence that previous welds were not entirely reproducible. In addition, the limited visual feedback system used previously has also led to inconsistent results within a single series (see Table 1), and substantial temperature variations within a single weld (see Fig. 4). By using temperature as a real time diagnostic, the surgical technique for vessel welding can be controlled, thereby resulting in more reproducible welds.

Although the surface temperature is both a convenient and important physical parameter to monitor for tissue welding, it does not provide a complete picture of the welding process. This is especially the case for thick vessels, where the heat is deposited superficially. In this case, a substantial temperature difference may exist between the outer adventitial layer (the temperature of which is monitored) and the inner elastin layers (the desired welding region). These thermal gradients can be calculated. Since the exposure times are generally long compared with the thermal diffusion times, a steady state solution should be valid for the welding process.

The results of this calculation are given in Figure 5 for two different types of tissue absorption depths, each at the same surface tissue temperature. The steady state solution was solved by using the formulas for heat conduction with convection at the boundaries and an exponential

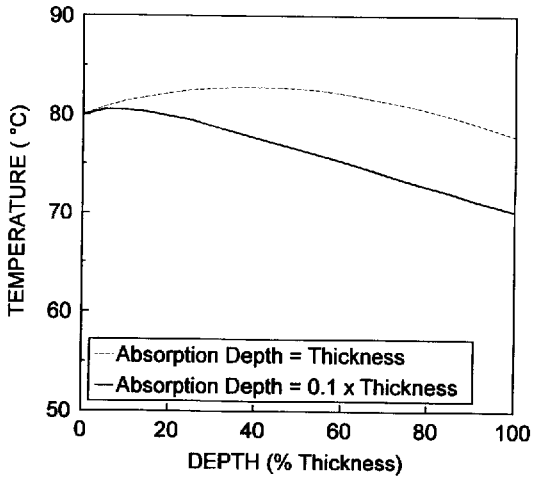

Fig. 5. Calculated temperature profiles vs. depth for laser heating a tissue surface to 80°C. The temperature across the tissue is shown for (**solid line**) a strongly absorbed laser, and (**dashed line**) the case where the absorption depth matches the tissue thickness. A steady state approximation was used.

heating term. For more details, see Springer and Welch [7]. As a result of heat loss at the surfaces, the maximum temperature always occurs in the interior of the tissue bulk. In one case, the absorption depth is much shorter than the vessel wall thickness. Here, as expected, a significant thermal gradient occurs across the vessel. The average temperature of the tissue is 76°C with a standard deviation of ±4°C for this low penetration depth case. This calculated temperature profile differs significantly from the 80°C surface temperature.

A second curve is shown for the case where the tissue absorption depth matches the vessel wall thickness. In this case, the temperature gradient across the tissue is significantly smaller. Here the average temperature is 81°C and the standard deviation is ±2°C. This average temperature is nearly the same as the surface temperature. Therefore, based on the calculated temperature profiles, if the absorption depth approximately matches the tissue thickness, the surface temperature should be a proper parameter to monitor the tissue welding process. In the studies mentioned earlier, the wavelength of the diode laser was chosen such that its absorption depth matches the vessel wall thickness, so the temperatures recorded should be indicative of the temperature at the weld site.

Although the use of infrared temperature measurements for tissue welding is relatively straightforward, several technical points should

be emphasized. First, only the temperature of the central portion of the laser heated spot should be imaged. This is necessary to minimize the effects of slight variations in the imaging angle, and, most importantly, to render the measurement insensitive to changes in the surrounding tissue temperature. Another thermal fluctuation that must be accounted for is the ambient "background" radiation seen by the detector (the imaging lens and the walls of the handle). In our thermometer, a thermistor was used to monitor the handle temperature and to provide gain compensation to accommodate a range of handle temperatures. Finally, since a single broadband detector was used, the temperature measurements were sensitive to the tissue emissivity, which may vary both for nonidentical vessels and for different sections of the same vessel. Based on the high water content in all vascular tissue, we assumed that the emissivity of the vessel was equal to that of water (0.99) and universal for all vessels. During welding, however, since desiccation probably occurs to some extent, the approximation of tissue as water may not be accurate. However, the organic contents of tissue are also expected to have emissivity near 1 in the infrared region detected by our thermometer, based on their high absorption in this region.

A straightforward solution to this problem caused by differences in the tissue emissivity is to use a two color detection scheme, where radiation over two separate regions of the infrared emission spectrum are measured. If the two wavelengths are properly selected, it is possible to determine the temperature independent of the tissue emissivity. We have demonstrated this over the temperature range of interest for tissue welding [12]; however, the extremely sensitive infrared detectors needed for this two color scheme are not amenable for use in in situ tissue welding.

Past tissue welding studies have indicated that laser assisted anastomoses are possible. These studies have utilized a plethora of laser wavelengths and several different surgical techniques, ranging from tissue cooling with a saline drip [2] to thermally activated glues and solders [13,14]. That all of these techniques have produced successful welds is both a testimony to the ingenuity of the research groups and a result of the relatively broad temperature range over which tissue fusion can occur, although consistent tissue welding can only be obtained over a more restricted temperature range. However, none of the previous studies have pointed to a particular system or technique positively suited for clinical use. Although speculative, one reason for the small clinical numbers may be the lack of consistency for successful vascular welds. The addition of real time diagnostics and closed loop controls, such as offered by thermally controlled systems, may provide the surgeon with the necessary tools to make tissue welding a clinical reality.

ACKNOWLEDGMENT

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